

# Detection of the tagged or untagged photons in acousto-optic imaging of thick highly scattering media by photorefractive adaptive holography

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We propose an original adaptive wavefront holographic setup based on the photorefractive effect (PR), to make real-time measurements of acousto-optic signals in thick scattering media, with a high flux collection at high rates for breast tumor detection. We describe here our present state of art and understanding on the problem of breast imaging with PR detection of the acousto-optic signal.

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## 1. Introduction

The field of acousto-optic imaging has been strongly stimulated by the deep and complete paper of W. Leutz and G. Maret<sup>1</sup>. In this paper the authors give a very clear view of the tricky interactions between light and sound in random media; this is why this work has stimulated a new active field now more "biomedical imaging" oriented and new detection schemes.

The present paper is at the frontier of two physical domains that are

1. detection of weak light signal by using photorefractive crystals,
2. breast cancer imaging by detection of the ultrasonic modulation of the light scattered through the breast.

Here, our purpose is to make a brief review of these two domains, and to describe the photorefractive detection of the scattered light modulated component in a pedagogical manner. By the way, we will describe our present state of art and understanding on the problem.

In this paper, we will first describe the basics principle of ultrasonic modulation of light imaging. We will, in

particular, introduce the concept of "ultrasonic tagged photons", which represents the weak signal to be detected. We will then describe how photorefractive adaptive holography can be used to detect the tagged photons. One must notice that all the groups working on the subject, exempt to us, do not detect the "tagged" photons, but "untagged" ones. The tagged photon signal is measured indirectly, since the total number of scattered photons (tagged + untagged) does not depend on the ultrasound. We will describe our technique and present our experimental results. In all these descriptions, we must not forget one difficulty that results from the decorrelation of the light that travels through breast organ. This effect known as speckle decorrelation is both due to the brownian motion of the scatterers, and to the breast inner motions (blood flow ...). In a typical in vivo situation, with 4cm breast thickness, the "speckle decorrelation time" is in the 0.1 to 1ms range. It is thus necessary to match the so called "photorefractive response time" with the "speckle decorrelation time". This effect, which is huge in breast, is not present in most of the ultrasonic modulation test experiments, which are performed with breast phantoms like dead tissues or diffusing gels. Since the decorrelation affects considerably the detection sensitivity, it is quite difficult to evaluate the figure of merit of the different techniques that are proposed to perform breast imaging. We will see that our setup, which is able to detect both the tagged and untagged signal, is also able to measure the photorefractive time *in situ*, i.e. with the same setup, same laser powers and same sample geometry than for breast imaging experiments. To our knowledge nobody is presently able to perform ultrasonic modulation imaging though 4cm of breast tissues in vivo. Experiments are under progress and we hope to be able to reach this aim in a near future.

## 2. Acousto-optic imaging

The combination of light and ultrasound to measure local optical properties through thick and highly scattering media is a tantalizing approach for *in vivo* imaging. It is an alternative solution to pure optical techniques for breast cancer detection. The use of light is motivated by its relative low absorption in the so called "optical therapeutic window" (700nm to 1000nm), and by the existence of optical contrasts between healthy and tumorous areas in this region of the spectrum.

Light is highly scattered within biological tissues, making direct optical study of thick sample very difficult to perform. Light scattering is characterized by two length parameters, *e.g* the scattering length  $l_s$ , and the light transport mean free path  $l_s^*$ . The scattering length  $l_s$  characterizes the memory of optical phase, and corresponds to the average distance that separates two scattering events. The light transport mean free path  $l_s^*$  characterizes the memory of the light propagation direction. In tissues,  $l_s$  is typically 50 to 100  $\mu m$ , while  $l_s^*$  is 10 $\times$  larger (0.5 to 1mm). Absorption of light is characterized by the absorption length  $l_a$ , which is in the 1 cm to 10 cm range. Absorption strongly depends on the nature of the tissue (optical contrast).

Because of scattering, direct imaging cannot be performed through more than a few millimeter thick samples. Contrarily to light, ultrasound (US) beams are ballistic in biological tissues. US gives thus access to millimeter range spatial resolution in thick sample (up to 4cm) yielding the development of the acousto-optic imaging that combines optics and ultrasound<sup>2,3</sup>.

### A. Principle: the tagged photons

Acousto-optic imaging is a hybrid technique, which combines, thanks to the acousto-optic effect, ultrasound and light. US are applied in the region of interest, within the thick scattering sample (see Fig.1 a). They make the scatterers vibrate. A CW laser (frequency  $\omega_L$ ) illuminates the sample. The vibration of the scatterers at the acoustic US frequency  $\omega_A$  (2 MHz typically) modulates the phase of the photons that are scattered by the sample. This is the so-called acousto-optic effect.

The light exiting the sample contains thus different frequency components (see Fig.1 b). The main component (the carrier) is centered at the laser frequency  $\omega_L$ . It is related to the diffused photons, that do not interact with the US. The sideband components are shifted by the US frequency  $\omega_L \pm \omega_A$ . The sideband photons, which result from the interaction between light and US, are called "tagged photons" (*i.e* photons tagged by the US).

The weight of the tagged photons components depends on the optical absorption in the region of interest, where the US beam is focused. Acousto-optics imaging stands in detecting selectively the tagged photons. An image of the sample optical absorption can be then built-up in scanning the US over the sample. Note that one of the difficulty in living tissues results from the motion

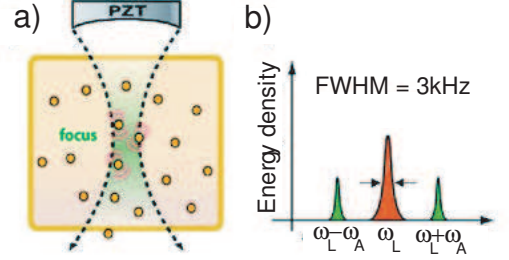


Fig. 1. Principle of acousto-optic imaging. a) Motion of the scatterer at  $\omega_A$ . b) Spectrum of the diffused light: carrier ( $\omega_L$ ), and sideband *i.e.* tagged photons ( $\omega_L \pm \omega_A$ ).

of scatterers (*e.g* blood flow) which broaden the carrier and sideband lines (see Fig.1 b). *In vivo* measurements through 4 cm breast tissues yield a broadening of 1.5kHz (Full Width at Half Maximum: FWHM)<sup>4,5</sup>.

### B. State of the art for the detection of the tagged photons

Many techniques have been proposed to detect the tagged photons. Marks et al.<sup>6</sup> investigated modulation of light in homogeneous scattering media with pulsed ultrasound. Wang et al.<sup>2,7</sup> performed ultrasound modulated optical tomography in scattering media. Lev et al. studied scattering media in the reflection configuration<sup>8</sup>. Wang and Shen<sup>9</sup> developed a frequency chirp technique to obtain scalable imaging resolution along the ultrasonic axis by use of a one-dimensional (1D) Fourier transform. Lev et al. use a set of optical fibers coupled to a single photodetector<sup>5,8,10</sup> that allows to work with samples, which decorrelate in time. Leveque et al.<sup>11,12,13</sup> performed parallel detection of multiple speckles on a video camera and demonstrated improvement of the detection signal-to-noise ratio of 1D images of biological tissues. The parallel detection has been still improved by Gross et al., which performs holographic detection reaching the shot noise sensitivity limit<sup>14</sup>, and by Atlan et al., which get resolution on the US propagation axis by using an holographic pulsed technique<sup>15</sup>.

All these methods exhibit two main limitations. First, the *optical etendue* (defined as the product of the detector area by the detector acceptance solid angle) of the detection system is not optimum, since it is much lower

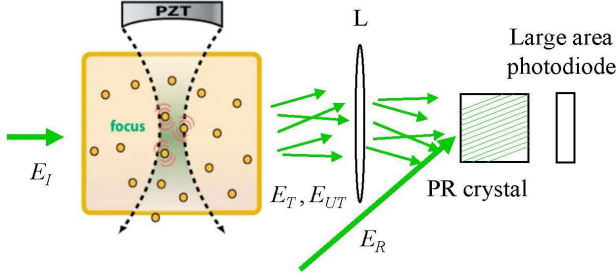


Fig. 2. Principle of the PR detection of the acousto-optic signal. PZT: ultrasound transducer,  $E_I$ : illumination optical field;  $E_T, E_{UT}$  tagged or untagged field scattered by the sample; PR crystal: photorefractive crystal; L: lens that collect the scattered field into the crystal and photodiode;  $E_R$ : PR crystal reference (or pump) field.

than the *etendue* of the tagged photons source. This *etendue* is the area of the sample (several  $\text{cm}^2$ )  $\times$  the emitting solid angle (which is about  $2\pi$  since the light is diffused by the sample in all direction). With a mono detector (photodiode)<sup>5,8,10</sup> the detection *etendue* is about  $\lambda^2$ . With a multi detector like a CCD camera<sup>11,12,13,14,15</sup> the *etendue* is  $N\lambda^2$ , where  $N$  is the is CCD number of pixel ( $N \sim 10^6$ ). Even with a camera, the *etendue* of detection is about  $\times 1000$  lower than the *etendue* of the emission.

The second problem occurs within living sample: the scatterers move, yielding in the frequency space a broadening of the tagged photons spectrum, as shown on Fig.1b<sup>4,5</sup>. This effect corresponds, in the time space, to a decorrelation of the tagged photons speckle pattern. Since all the methods described above perform coherent detection, the bandwidth of detection is limited by the detector bandwidth. With camera (there is no problem of bandwidth with photodiode, but the *etendue* is much lower), the bandwidth is roughly equal to the camera image frequency  $\omega_{CCD}$ , which is in general much lower ( $\omega_{CCD} \sim 10 \dots 100$  Hz) than the tissue broadening (3 kHz). It is still possible to work with fast camera (kHz), but in that case i) the camera quantum efficiency is lower (CMOS), and ii) the number of pixel  $N$  is limited, because  $N \times \omega_{CCD}$  is the flux of information to transfer to the computer, and this flux is limited ( $< 10^6 \dots 10^7 \text{ s}^{-1}$ ).

### 3. The photorefractive (PR) detection of the acousto optic signal

More recently has appeared a new tagged photons detection technique that is based on the photorefractive effect (PR) and that is illustrated by Fig.2. The light that is scattered by the sample ( $E_T$  or  $E_{UT}$  for the tagged or untagged field) is detected by a photorefractive detector (PR crystal + photodiode PD) that is pumped by a reference field  $E_R$ .

Since the crystal and the photodiode might be quite

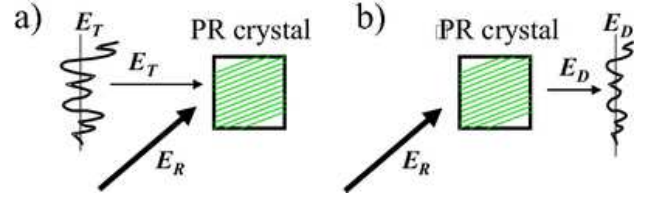


Fig. 3. Principle of holography using the photorefractive effect.

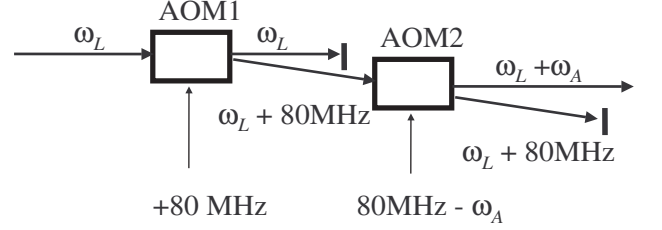


Fig. 4. Method for tuning the frequency  $\omega_R$  of the reference beam:  $\omega_R = \omega_L + \omega_A$  for example. AOM1 and AOM2 are acousto optic modulators.

large (up to  $1\text{cm}^2$ ) and since the light is collected by a large Numerical Aperture (N.A.  $\sim 1$ ) collecting lens, the photorefractive detection benefits of a high *etendue*, about  $100\times$  larger than in a typical camera with  $N \sim 10^6$  pixels. We will see that the detection bandwidth is the inverse of the "photorefractive time"  $T_{PR}$ . We get for example, for a  $1\text{W}/\text{cm}^2$  pump beam,  $1/T_{PR} \sim 1 \text{ kHz}$ <sup>16,17</sup>. This bandwidth, which is about  $100\times$  larger than for a  $N \sim 10^6$  CCD camera<sup>1</sup>, is within the range of the linewidth ( $\sim 3 \text{ kHz}$ ) of the light scattered *in vivo* by a breast organ<sup>4</sup>.

#### A. The volume hologram

Photorefractive effect arises in materials that present both electrooptic effect and photoconductivity, which combination allows to transform a non uniform illumination of the material into a spatial variation of the refractive index<sup>18</sup>. When illuminated by the interference pattern between an object and a reference beam, the material records a hologram, *i.e* the amplitude and phase of the object beam. This hologram is dynamic meaning that it can follow the interference pattern fluctuations slower than the response time  $T_{PR}$  of the material, also meaning that only slowly moving hologram are recorded.

The first effect is the recording of the signal beam information (phase and amplitude of the signal field  $E_T$ )

<sup>1</sup> we implicitly exclude here fast CMOS camera because of poor quantum efficiency and noise, and because of finite bandwidth for the data transfert from camera to computer

within the PR crystal under the form of local changes of the crystal refractive index  $n$  yielding a volume hologram (Fig.3a). Since the recording takes a finite time  $T_{PR}$  (0.1 to 10ms in our device), the PR effect selects, within the signal beam, the components whose frequency is close or equal to the reference beam frequency  $\omega_R$ .

Here, the large, almost flat field, reference beam (field  $E_R$ ) interferes with signal field  $E_T$  onto the PR crystal. In order to select the tagged or untagged photons, the frequency  $\omega_R$  of the reference beam is made equal to the tagged or untagged photons frequency:  $\omega_R = \omega_L \pm \omega_A$  or  $\omega_R = \omega_L$  respectively. To adjust  $\omega_R$  one can use for example two acousto-optic modulators AOM1 and AOM2 (Bragg cells) as shown on Fig.4. With such a choice, the interference pattern  $E_T E_R^*$  of the reference beam with the selected photons beam varies slowly in time. The selected beam information can thus be grooved within the PR crystal volume hologram.

In the case of a perfect monochromatic signal beam, the local variation of the hologram refractive index  $\delta n$  is simply proportional to the modulation depth of the interference pattern  $(E_T E_R^*) / (|E_T|^2 + |E_R|^2)$ . If a time modulation is added on the signal (e.g. amplitude or phase modulation), we have to take into account the finite time  $T_{PR}$  needed to groove the hologram, and we get<sup>19,20</sup>:

$$\delta n \propto \frac{\langle E_T E_R^* \rangle_{T_{PR}}}{|E_T|^2 + |E_R|^2} \quad (1)$$

where  $\langle \rangle_{T_{PR}}$  is the average over the grooving time  $T_{PR}$ , average which is defined by:

$$\langle A \rangle_{T_{PR}} = \frac{1}{T_{PR}} \int_0^\infty A(t - \tau) e^{-\tau/T_{PR}} d\tau \quad (2)$$

### B. The diffracted beam $E_D$

The second effect is illustrated by Fig.3b. The reference beam ( $E_R$ ) is diffracted by the volume hologram yielding a diffracted beam ( $E_D$ ). The diffracted field  $E_D$  is simply proportional to the hologram refractive index changes  $\delta n$  and to the reference beam field  $E_R$ . We get thus:

$$E_D \propto \frac{\langle E_T E_R^* \rangle_{T_{PR}}}{|E_T|^2 + |E_R|^2} E_R \quad (3)$$

In typical application the reference beam intensity is much larger than the signal beam one, and except of the average over  $T_{PR}$ ,  $E_R$  and  $E_R^*$  simplifies in Eq.4 yielding  $E_D \propto E_T$  i.e.

$$E_D \simeq \eta E_T \quad (4)$$

where  $\eta = 0.1..0.5$  is a numerical factor which mainly depends on the crystal.

Eq.4 is valid, when the decorrelation of the signal field  $E_T$  can be neglected during the grooving time  $T_{PR}$ , i.e. when

$$\delta \omega T_{PR} \ll 1 \quad (5)$$

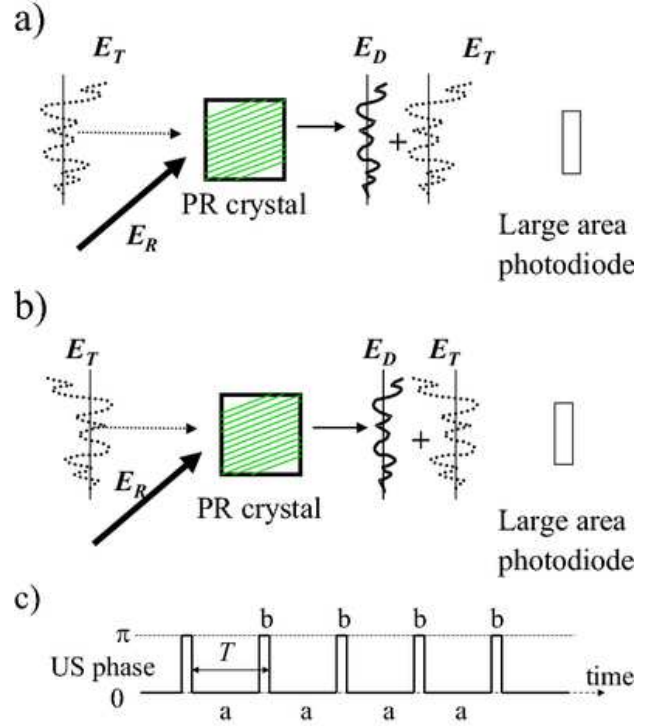


Fig. 5. a,b) Detection of the tagged photons, when the phase of US beam is zero (a) and  $\pi$  (b). c) Phase of the US beam.

where  $\delta \omega$  is the frequency width of the signal beam ( $\Delta \omega \sim 3$  kHz for the breast).

We have to notice that an increase of the reference beam intensity  $|E_R|^2$  does not change  $\eta$ , but reduces the grooving time  $T_{PR}$ , since  $T_{PR} \propto 1/|E_R|^2$ . The main advantage of increasing the reference beam power  $|E_R|^2$  is thus to reduce  $T_{PR}$  enough to neglect the signal field decorrelation. Condition of Eq.5 is then fulfilled, and the Eq.4 limit can be reached.

Since the volume hologram has recorded the mode structure of the signal beam versus reference beam interference pattern, and since the pump beam is diffracted by the hologram, the diffracted beam ( $E_D$ ) has the same mode structure than the signal beam ( $E_T$ ) (see Eq.4). This result is illustrated by Fig.3b where  $E_D$  is displayed with the same shape than  $E_T$  on Fig.3a, but with a smaller amplitude ( $\eta < 1$ ).

The signal ( $E_T$ ) and diffracted ( $E_D$ ) beams are thus spatially coherent. They can interfere constructively (or destructively) on a large area ( $\sim 1$  cm<sup>2</sup>) light mono detector (i.e. a photodiode). This property will be useful to detect efficiently the tagged and untagged photons signal.

### C. Detection of the tagged photons

The principle of tagged photons detection is illustrated by Fig.5. The phase  $\varphi$  of the US beam is reversed periodically ( $\varphi = 0$  or  $\pi$ ) with period  $T$  (see Fig.5c). The

phase of the tagged photons field  $E_T$ , which follows the US phase, is then reversed too.

To simplify the discussion, we will neglect the decorrelation of the tagged photons field (Eq.5 is fulfilled). We will also modulate the phase rapidly (with respect to  $T_{PR}$  i.e. with  $T \ll T_{PR}$ ), keeping  $\varphi$  zero most of the time (see Fig.5 c), so that the hologram can be considered as static and unperturbed by the phase variation.

In that case, the diffracted field  $E_D$  will remain nearly constant:  $E_D \simeq \eta \langle E_T \rangle$ . When  $\varphi$  is zero,  $E_T$  and  $E_D$  are in phase, they interfere constructively and the total intensity signal  $|E_T + E_D|^2$  is maximum (see Fig.5a):

$$I_0 = (|E_T + E_D|^2)_{\varphi=0} \simeq |E_T|^2(1 + \eta)^2 \quad (6)$$

When phase is  $\pi$  contrarily, the  $E_T$  and  $E_D$  are opposite in phase, and the total intensity signal is minimum (see Fig.5b):

$$I_\pi = (|E_T + E_D|^2)_{\varphi=\pi} \simeq |E_T|^2(1 - \eta)^2 \quad (7)$$

Reversing the phase of the US yields to a modulation of the total intensity signal equal to :

$$I_0 - I_\pi \simeq 4 \eta |E_T|^2 + \dots \quad (8)$$

#### D. Detection of the untagged photons

It is a little bit more difficult to illustrate the detection of the untagged photons by a simple figure, because the calculation involves to consider both the untagged photons field at the carrier frequency  $\omega_L$ , and the tagged photons fields  $E_T$  and  $E_{T'}$  which evolves at the two sideband frequencies  $\omega_L + \omega_A$  for  $E_T$ , and  $\omega_L - \omega_A$  for  $E_{T'}$ . To detect the untagged photons, we tune the reference beam frequency  $\omega_R$  at the untagged photons frequency:  $\omega_R = \omega_L$ , and we modulate the US beam intensity by turning on and off the US beam.

To simplify the discussion, we will neglect again the decorrelation of the tagged photons field (Eq.5 is fulfilled). We will also modulate the US beam rapidly (with respect to  $T_{PR}$  i.e. with a period  $T \ll T_{PR}$ ). Let us call  $E_U$  and  $E_{U'}$  the untagged photons fields without, and with US beam.  $E_T$  and  $E_{T'}$  are the tagged photons fields with US (theses fields are zero without US).

Since the energy is conserved, the total number of photons (carrier + sidebands) does not depend on the US. We get thus:

$$|E_U|^2 = |E_{U'}|^2 + |E_T|^2 + |E_{T'}|^2 \quad (9)$$

The untagged photons field in presence of the US, e.g  $E_{U'}$ , is spatially coherent with the one without US, e.g  $E_U$ . According to (Eq.9), its magnitude can be expressed as follows :

$$|E_{U'}| = |E_U| \sqrt{1 - \frac{|E_T|^2 + |E_{T'}|^2}{|E_U|^2}} \quad (10)$$

In practical situation, the efficiency of the acousto optic effect is low and the energy within the sideband is low

(< 1%) with respect to the carrier. This means that the untagged photons field variation is low:  $E_U - E_{U'} \ll E_U$ . Whatever the value of the cyclic ratio modulation is, one can thus consider that the PR effect involves  $E_U$  only. We get:

$$E_D \simeq \eta E_U \quad (11)$$

When the US is off, the field on the detector is  $E_U + E_D$  and the detected intensity signal  $I$  is:

$$I = |E_U + E_D|^2 \quad (12)$$

$$= |E_U|^2 + |E_D|^2 + 2\eta |E_U|^2 \quad (13)$$

When the US is on, the field on the detector is  $E_{U'} + E_D$  for the carrier, and  $E_T$  and  $E_{T'}$  for the two sidebands. The intensity signal  $I'$  is:

$$I' = |E_{U'} + E_D|^2 + |E_T|^2 + |E_{T'}|^2 \quad (14)$$

$$= |E_{U'}|^2 + |E_D|^2 + \eta(E_{U'}E_U^* + E_U E_{U'}^*) + |E_T|^2 + |E_{T'}|^2 \quad (15)$$

Taking into account the energy conservation (Eq.9), the spatial coherence of  $(E_U, E_{U'})$  and (Eq.10), we get the modulation of the detected intensity:

$$I - I' = 2\eta (|E_U|^2 - |E_{U'}||E_U|) \quad (16)$$

$$\simeq 2\eta \frac{|E_T|^2 + |E_{T'}|^2}{2} \simeq 2\eta |E_T|^2 \quad (17)$$

since the weight of the two sidebands components are approximately the same:  $|E_T|^2 \simeq |E_{T'}|^2$ .

By comparing Eq.8 and Eq.16, the detected signal have the same order of magnitude when detecting either the tagged or the untagged photons, when we consider the same acoustical energy.

#### E. Detecting tagged or untagged photons ?

To our knowledge, three groups are working on acousto optic imaging with PR detection of the signal. Two of them, the R.A. Roy<sup>17,21,22,23</sup> and the L.V. Wang group<sup>24</sup> detect the untagged photons. We are the third group<sup>16,25</sup> and we detect both the tagged and untagged photons.

Detection of the untagged photons is simpler since it is not necessary to shift frequency the reference beam ( $\omega_R = \omega_L$ ). The acousto optic modulators of Fig.4 are thus not needed. Moreover, it is not necessary to apply the US beam all the time. Untagged photons detection is thus well suited to detect very short burst of US beam able to give information resolved along the US beam propagation direction<sup>17</sup>. But short US burst yield a small signal, and signal is needed to image thick breast *in vivo*.

The detection of the untagged photons corresponds to a small change on a large signal (white background detection), while the detection of the tagged photons, which corresponds to roughly the same absolute value change, yields contrarily to about 100% change on a small signal (black background detection). Tagged photon detection



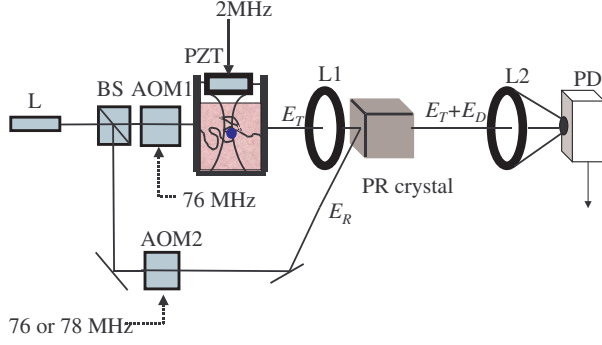


Fig. 6. Typical experimental setup. L: laser; BS: Beam Splitter; AOM1, AOM2: acousto optic modulators; PZT: US generator; L1, L2: light collecting lenses; PR: photorefractive crystal; PD: photo detector.

is thus expected to give less technical noise. For example, vibrations on the reference beam mirrors, which modify the length of the pump beam arm, is expected to yield about  $100 \times 2$  more noise for the untagged configuration than for the tagged one.

The tagged photon configuration offers more degrees of freedom for the detection configuration, because the signal and reference beam can be modulated whether in phase or amplitude.

Since we do not have tested all the possible detection configurations, making a complete comparison of tagged and untagged photons detection schemes is out of the scope of the present paper. We can simply say that for the configurations we have presently tested, the Signal to Noise Ratio (SNR) is about the same in the two cases. Since our purpose is to image breast, we need to improve the detection sensitivity. We continue thus to work with our setup that is able to detect both tagged and untagged photons, exploring configurations that are expected to yield better SNR. This work is under progress.

#### 4. Experimental test

##### A. Setup

A typical experimental setup, able to select either the tagged or the untagged photons, is shown on Fig.6. The main laser beam is splitted in an illumination and reference beam by the beam splitter BS. The US beam (2MHz) produced by the generator PZT is focused within the sample. The frequency offset of the illumination beam is adjusted by using the two acousto-optic modulators AOM1 and AOM2 which are excited at 78MHz and 76MHz (for selecting the tagged photons), or both 78MHz (for selecting the untagged pho-

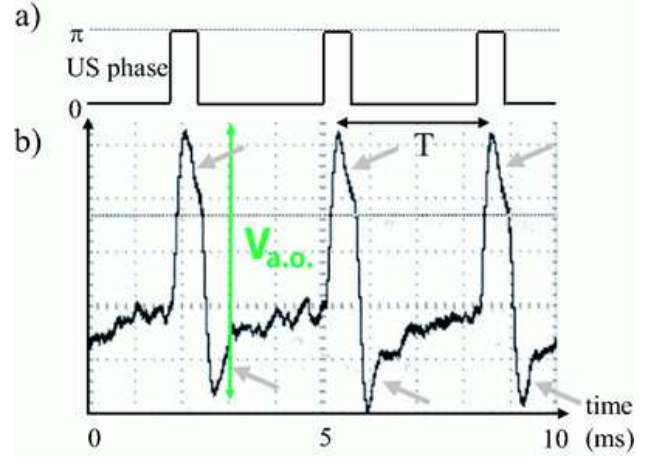


Fig. 7. Typical tagged photons experimental signal. a) Modulation of the US phase b) Tagged photons signal.  $V_{a.o.}$ : maximum variation of the tagged photons signal. T: phase modulation period. Vertical axis is voltage at the output of the PD amplifier: 100mV per div.

tons). The light diffused by the sample is collected by the high  $NA \sim 1$  (Numerical Aperture) lenses L1 and L2. L1 collects the light within the PR crystal that records the hologram of the selected signal beam (tagged or untagged). L2 collects the interference pattern of the signal beam ( $E_T$ ) with the diffracted beam ( $E_D$ ) into the photodetector PD.

In our setup, L is a Nd:YAG laser (1.06  $\mu m$ , 1 to 5 W CW power), the PR crystal is a  $1.4 \times 1.4 \times 2 cm^3$  GaAs crystal<sup>26</sup>, and PD is a large area photodiode (0.1 to 0.5  $cm^2$ ) whose signal is amplified by a transimpedance amplifier ( $R = 100 K\Omega$  to 10 M $\Omega$ ).

In the Murray's setup<sup>17</sup>, L is a frequency doubled Nd:YAG (532 nm, 80 mW), the PR crystal is a  $5 \times 5 \times 7 mm^3$  Bi<sub>12</sub>SiO<sub>20</sub> crystal, whose PR efficiency is improved by applying a DC electric field, the US frequency is 1.1 MHz, and PD is an avalanche photodiode. Since Murray detects the untagged photons, the acousto-optic modulators are not present, but it should be pointed out that absorption at 532nm is more important than at 1064nm, and thus it can reduce the thickness of investigation.

Our setup, which can detect both the tagged and untagged photons, is expected to be more sensitive, while the Murray's setup, which is used with short US pulses, is faster.

##### B. Experimental result

Fig.7 shows a typical tagged photons experimental signal obtained with 0 to  $\pi$  phase modulation of the US beam. The modulation frequency is 300Hz (modulation period  $T = 3.33$  ms). The US beam frequency is 2MHz, with a maximum US pressure of 2MPa at the US beam waist. The main laser power is 1.2W. The reference and illumination beam power are both 300mW, their areas on

<sup>2</sup> here 100 is the untagged versus tagged photons field ratio

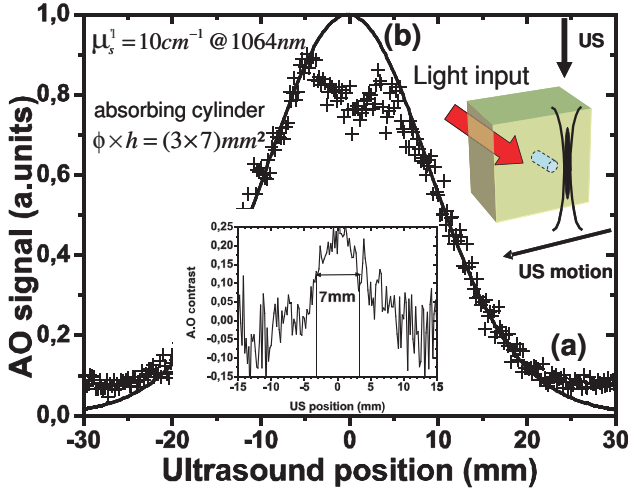


Fig. 8. Acousto-optic profile (a) with an optical absorber embedded within an Agar + *Intralipid* phantom of thickness  $t = 30\text{mm}$  with a reduced scattering coefficient  $\mu'_s = 10\text{cm}^{-1}$  @1064nm obtained with an anisotropic (PR) detection configuration. Light input on the sample is  $1\text{W/cm}$  (gaussian illumination  $FWHM = 1.1\text{mm}$ ). The US pressure ( $2.3\text{MHz}$ ,  $1.5\text{MPa}$ ) is phase modulated at  $3\text{kHz}$  with a duty cycle of 24%. Curve (b) represents a fit of the one dimensional extend of light scattering (e.g  $\exp(-\mu_{eff}r)/r$  with  $\mu_{eff} = 2.2\text{cm}^{-1}$ ) within the US plane (15mm from input window), weighted by the input illumination  $\mu_{eff} = 2.2\text{cm}^{-1}$ . Inset represents the acousto-optic contrast  $[1 - (a)/(b)]$ .

crystal and sample are both  $1\text{cm}^2$ .

Measurement is performed with a  $4\text{cm}$  chicken sample, whose optical properties (diffusion and absorption) are close to human breast. As seen, the tagged photons signal SNR is good (16 times averaged). One must notice that the signal is not rectangular like the phase modulation. In particular, the maximum of signal, which occurs on the phase plateaus ( $\varphi = \pi$  and  $0$ ) is not flat, but decreases exponentially (see grey arrows on Fig.7).

This is expected when the PR time  $T_{PR}$  becomes shorter than the phase modulation period  $T$ . We have measured (see further)  $T_{PR}$  and we have found  $T_{PR} = 0.5$  ms. Note that  $T_{PR}$  can also be measured on Fig.7, since  $T_{PR}$  is time constant of the grey arrow decay.

This result is very encouraging, because it means that the detection bandwidth is  $1/(2\pi T_{PR}) = 0.3\text{kHz}$ . Remember the signal bandwidth is  $\Delta\omega = 1.5\text{kHz}$  (HWHM) on the breast. The detection is thus optimal, within a factor 5. Since the SNR is very high (much larger than 5) in Fig.7, we expect to get enough SNR to get significant result with a thick living sample.

As an other illustration, Fig.8(a) represents a profile of a Agar plus *Intralipid* phantom with a thickness  $t = 30\text{mm}$  and a reduced scattering coefficient  $\mu'_s = 1/l_s^* = 10\text{cm}^{-1}$  at  $1064\text{nm}$ . The absorption co-

efficient of water @1064nm is  $\mu_a = 0.144\text{cm}^{-1}$ <sup>27</sup>. The sample contains an optical absorber (black ink), which is a cylinder with a diameter of  $3\text{mm}$  and a height of  $7\text{mm}$  along the laser input direction (perpendicular to the US beam). The magnitude of the US pressure is approximately of  $1.5\text{MPa}$  at  $2.3\text{MHz}$ , with a phase modulation at  $3\text{kHz}$  and a duty cycle of 24%, corresponding to the maximum of the signal with a lock-in detection<sup>16,25</sup>. The photorefractive holographic setup is based on an anisotropic diffraction configuration<sup>28</sup>: the reference beam (e.g vertically polarized) diffracts a contribution (e.g tagged-photons field) which is perpendicularly polarized (e.g horizontal); the output speckle from the sample is  $45^\circ$ -polarized from vertical direction using a large aperture infrared dichroic polarizer, and a similar analyzer is positioned in front of the photodetector with an horizontal polarization axis. Consequently, the speckle and the reference fields still interfere within the PR crystal in order to build the hologram, the diffracted reference and the speckle field recombine onto the analyzer as well. This configuration minimizes the collection of the unwanted scattered reference by the PR crystal faces. In this experiment the tagged light is about  $\times 10^4$  lower than the total scattered light (untagged photons plus scattered reference light).

Classically, in the 3D diffusion regime and in presence of absorption, the spatial distribution of energy emitted from a point source at distance  $r$  is given by  $\frac{1}{r} e^{-\mu_{eff}r}$ , where  $\mu_{eff} = \sqrt{3\mu_a(\mu_a + \mu'_s)}$ . This effective parameter indicates that attenuation is increased by scattering, that lengthens optical pathes.

The continuous envelope Fig.8(b) represents the fit the experimental data Fig.8(a) using this model and taking into account the gaussian input illumination ( $FWHM = 1.1\text{mm}$ ). The effective coefficient  $\mu_{eff}$  is founded to be  $2.2\text{cm}^{-1}$ , close to the theoretical value ( $\mu_{eff} = 2.1\text{cm}^{-1}$ ) given by the reduced scattering coefficient of the medium and the absorption coefficient of pure water at  $1064\text{nm}$  defined above. The measured background (around  $0.8\text{mV}$ ) corresponds to the noise of the transimpedance stage of the detection, that is shot-noise limited at this level of the scattered light. The absorbing element is revealed by the acousto-optic contrast, e.g  $[1 - (a)/(b)]$ , which is close to 0.22, and exhibits a  $FWHM$  of  $7\text{mm}$ . This value is connected to the diameter of the absorbing element ( $3\text{mm}$ ), the US resolution (just above  $1.5\text{mm}$ ) and the light transport mean free path  $l_s^*$  of the scattering medium (about  $1\text{mm}$ ).

## 5. Measurement of the photorefractive time $T_{PR}$

Most published results on ultrasound light modulation imaging have been obtained with phantoms, which do not decorrelate in time. In that case, the PR detection SNR does not strongly depends on the reference beam power. The power must be large enough to reach the plateau value for the photorefractive efficiency  $\eta$ , but remains low enough to avoid noise (the reference beam is

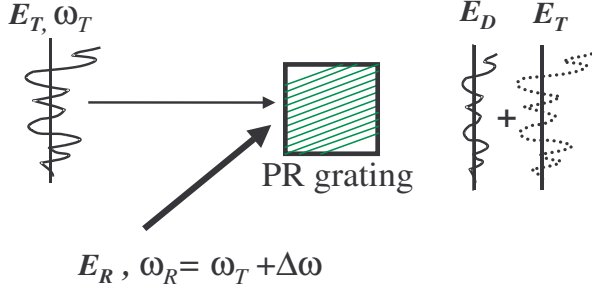


Fig. 9. PR effect when the selected photons and reference beam frequencies are different:  $\omega_R \neq \omega_T$

scattered by the PR crystal defects yielding a parasitic photodiode current that brings noise). With phantoms, the best sensitivity is then obtained with a quite low power reference beam ( $< 100$  mW in our experiment).

With breast, the light signal is Doppler broadened by the tissues inner motions (brownian motion, blood flow...) yielding typically a spectral width of  $3\text{kHz}$ . In order to optimize the detection efficiency, one must increase the detection bandwidth  $1/T_{PR}$  by increasing the reference beam power to obtain  $1/T_{PR} \sim 3\text{kHz}$ . Optimal detection conditions for phantoms and breast are thus very different.

To improve the detection sensitivity for future breast experiment, it is very important to measure  $T_{PR}$ . To get reliable result, we have proposed a technique able to measure  $T_{PR}$  *in situ*, i.e. in the setup that is used for imaging phantoms (and breast in future)<sup>29</sup>.

#### A. Principle of the measurement of $T_{PR}$

To perform acousto-optic imaging with living sample, it is very important to measure  $T_{PR}$ , since it is necessary to match  $T_{PR}$  with the sample decorrelation time in order to optimize the detection efficiency. The ability in our setup to freely choose the frequency of the reference beam gives new opportunity to measure the photorefractive time  $T_{PR}$ .

The idea of the measurement technique is illustrated by Fig.9. The frequency of the reference beam  $\omega_R$  is shifted with respect to the signal beam frequency  $\omega_T = \omega_L + \omega_A$  or  $\omega_L$ :

$$\omega_R = \omega_T + \Delta\omega \quad (18)$$

To simplify the notation, we will consider that  $E_R$  still represents a complex field projection at frequency  $\omega_T$ , so that  $E_R$  must be replaced in the calculation by  $E_R e^{j\Delta\omega t}$ . We get thus:

$$\delta n(t) \propto \frac{\langle E_T E_R^* e^{-j\Delta\omega t} \rangle_{T_{PR}}}{|E_T|^2 + |E_R|^2} \quad (19)$$

and

$$E_D(t) \propto \frac{E_R e^{j\Delta\omega t}}{T_{PR}} \frac{\int_0^\infty E_T(t-\tau) E_R^* e^{-j\Delta\omega\tau} e^{-\tau/T_{PR}} d\tau}{|E_T|^2 + |E_R|^2} \quad (20)$$

Note that if  $\Delta\omega$  is zero, Eq.20 is identical to Eq.3. Note also that  $E_R^*$  does not depend on time, and can thus be removed from the integral.

Consider that a PR experiment is made with a sample whose decorrelation time is much longer than  $T_{PR}$ . This means that decorrelation can be neglected, and that  $E_T(t)$  is uniquely driven by the US amplitude or phase modulation.  $E_T(t)$  is thus known.  $|E_T|^2$  can also be neglected in Eq.20 denominator (since  $|E_T|^2 \ll |E_R|^2$ ) and thus :

$$E_D(t) \propto \frac{e^{j\Delta\omega t}}{T_{PR}} \int_0^\infty E_T(t-\tau) e^{-j\Delta\omega\tau} e^{-\tau/T_{PR}} d\tau e^{j\Delta\omega t} \quad (21)$$

We must notice that in Eq.21,  $E_T(t)$  is convolved by two time kernels. The first kernel  $e^{-\tau/T_{PR}}$  is unknown (since  $T_{PR}$  is unknown), while the second  $e^{-j\Delta\omega\tau}$  is known. Its width can be freely adjusted by tuning  $\Delta\omega$  with the acousto-optic modulator.

From Eq.20 it is then straightforward to calculate the dependance of the acousto-optic signal with  $\Delta\omega$  for the different detection configurations (phase modulation for the tagged photons and amplitude modulation for the untagged photons)<sup>29</sup>. Comparing the calculated spectrum to the experiment yields then an accurate measurement of  $T_{PR}$ .

#### B. Calculation of the tagged photons signal with US amplitude modulation

It is possible to calculate the tagged photons signal as a function of  $\Delta\omega$  in the phase modulation configuration. Nevertheless, as shown in<sup>29</sup>, the shape of the spectrum is quite cumbersome, and it seems quite heavy to fit the experimental data with such a spectrum shape.

It is thus more efficient to measure  $T_{PR}$  with a rectangular amplitude modulation of the US with 50% cycling ratio, the tagged (or untagged) acousto-optic signal being measured with a lock-in amplifier tuned at the modulation frequency. This is the key point of the detuning method, since measurements are performed *in situ* at the US modulation frequency (here  $2.5\text{kHz}$ ) and thus do not depend on the frequency response of the detector, which is quite distorted due to the many stages of electronic filters connected to the photodetector. A straightforward calculation gives only three contributions for the  $P$  and  $Q$  quadrature of the lock-in signal<sup>29</sup> :

$$P = P_0 + P_+ + P_- \quad (22)$$

$$Q = Q_0 + Q_+ + Q_- \quad (23)$$

with

$$P_0(\Delta\omega) = \frac{2A}{1 + (\Delta\omega T_{PR})^2} \quad (24)$$



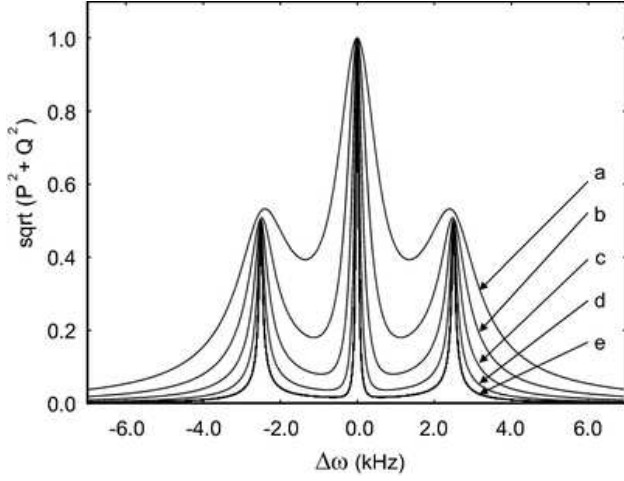


Fig. 10. Calculated spectrum of the tagged photons signal (amplitude modulation with 50% cycling ratio) with  $T_{PR} = 0.25ms$  (a),  $0.5ms$  (b),  $1ms$  (c),  $2ms$  (d),  $4ms$  (e). Horizontal axis is the frequency offset  $\Delta\omega$ . Vertical axis is  $\sqrt{P^2 + Q^2}$  lock-in signal in arbitrary normalized units.

$$P_{\pm}(\Delta\omega) = \frac{A}{1 + (\omega_{mod} \mp \Delta\omega)^2 T_{PR}^2} \quad (25)$$

$$Q_{\pm}(\Delta\omega) = -\frac{A(\omega_{mod} \mp \Delta\omega)T_{PR}}{1 + (\omega_{mod} \mp \Delta\omega)^2 T_{PR}^2} \quad (26)$$

where  $A$  is a proportional constant. By using Eq.24 to Eq.26 it is then quite simple to measure  $T_{PR}$  by fitting the experimental data with the calculated  $\Delta\omega$  spectrum.

### C. Measurement with the tagged photons and US amplitude modulation

From Eq.22 to Eq.26 we have calculated the tagged photons signal as a function of  $\Delta\omega$ , when the US beam is modulated with a rectangular  $[0, 1]$  amplitude modulation of 50% cycling ratio, the detection being performed with a lock amplifier tuned at the modulation frequency ( $2.5kHz$ ). As seen on Fig.10 the shape of the spectrum is strongly dependent on  $T_{PR}$ .

It is then possible to fit experimental data on the theoretical curves. Fig.11 shows the magnitude (e.g.  $R = \sqrt{P^2 + Q^2}$ ) of the lock-in signal (points). The tagged photons are selected ( $\omega_R = \omega_L + \Delta\omega$ ) and the US beam is modulated in amplitude with 50% duty cycle. The reference beam flux of  $300mW/cm^2$ , and the modulation frequency is  $2.5kHz$ . We have fit the experimental data with the theoretical curve deduced from Eq.24 to Eq.26. The fit free parameters are  $T_{PR}$  and  $A$ . The best fit yields  $T_{PR} = 0.45ms$ . The experimental data are on Fig.11 as a black curve, the fit as an heavy grey line curve.

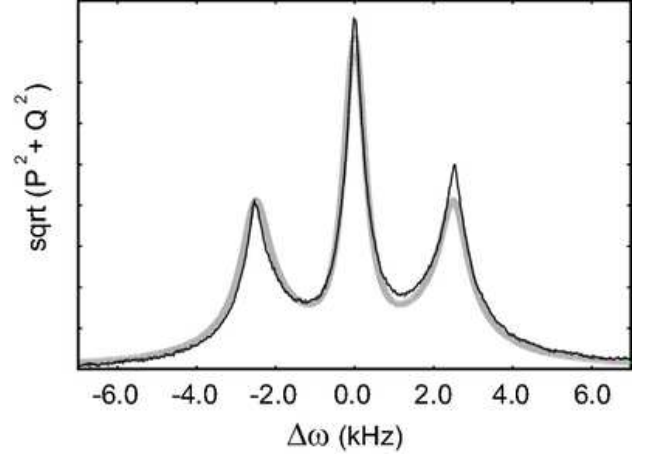


Fig. 11. Spectrum of the tagged photons signal with 50% duty cycle amplitude modulation of the US. Black curve is experimental data with reference beam flux of  $300mW/cm^2$  and a modulation frequency of  $2.5kHz$ . Heavy grey line curve is calculated spectrum with  $T_{PR} = 0.45ms$ . Horizontal axis is the frequency offset  $\Delta\omega$ . Vertical axis is  $\sqrt{P^2 + Q^2}$  lock-in signal in arbitrary units.

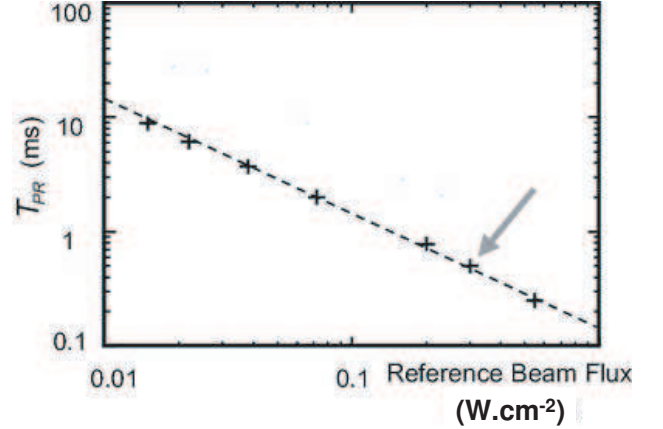


Fig. 12.  $T_{PR}$  in  $ms$  as a function of the reference beam flux in  $W/cm^2$ . Crosses are experimental data. Dashed line is data extrapolation with slope  $-1$ . Grey arrow corresponds to Fig.7 experimental conditions.

We have recorded many spectra for different reference beam flux. Each spectrum has been fitted by the theoretical curve yielding  $T_{PR}$ . Fig.12 shows in log-log scale  $T_{PR}$  as a function of the reference beam flux. Experimental points are crosses, data linear log-log extrapolation is dashed line. As seen, the slope of the extrapolation line is  $-1$ . This means that  $T_{PR}$  is inversely proportional to the beam flux, as expected.

The shortest photorefractive time we get is  $T_{PR} = 0.25ms$  for a flux of  $0.55 W/cm^2$ . The Fig.7 modulation phase signal is obtained with  $T_{PR} = 0.5ms$  and

$0.3W/cm^2$  (grey arrow on Fig.7).

## 6. Conclusion

Seeing through highly scattering media such as living tissues is a goal difficult to reach. Coupling light and ultrasound in acousto-optic imaging is a promising method to reach this aim. Nevertheless the efficient detection of the tagged photons remains a challenge.

The PR crystal detection scheme proposed here is a possible way to solve this problem. PR crystal detection has many advantages. The detection optical etendue is large since photodetector area may be quite large ( $\sim 1cm^2$ ), and since the collecting lens numerical aperture can be large, too ( $NA \sim 1$ ). Since the detector is a single-detector (photodiode), the analysis of the data is simple and fast. By adjusting the power of the pump beam, it is possible to match the detection bandwidth  $1/T_{PR}$  with the signal bandwidth  $\Delta\omega$  in order to detect with optimal efficiency the "tagged" or "untagged" signal diffused by living tissues that are broadened by the diffuser inner motion (brownian motion, blood flow ...). We demonstrate here our ability to get a high SNR (see Fig.7) with a thick chicken sample. Our chicken sample does not decorrelate as do living tissues, the measurement is done with a short photorefractive time  $T_{PR} = 0.5$  ms. This result is very encouraging.

The results presented in this paper have been obtained with a Nd:YAG laser at 1064 nm and a GaAs photorefractive crystal. The method could be significantly enhanced by the use of a laser source at 800 nm, according to the absorption coefficients of hemoglobin and deoxyhemoglobin, in order to perform a measurement of the local blood activity (two wavelengths measurements). We are searching at present for new PR crystals that are sensitive in this spectral range.

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